Ultrasound for Interventional Pain Management

An Illustrated Procedural Guide

Philip Peng
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This book is dedicated to my wife, Carol, for her continued support, encouragement, and understanding;
to my children, Julia and Michael, who fill me with joy and love;
and to my sister, Rita, who keeps reminding me to be strong and assertive.
Without them, this book would be possible.

Philip Peng

Preface

In the last 15 years, we witnessed a rapid surge in interest in applying ultrasound-guided pain intervention. Before 2003, the interest in ultrasound-guided pain intervention was mostly restricted to musculoskeletal system. Since then, many new techniques in ultrasound-guided pain intervention were developed in various peripheral and axial structures among pain specialists. More recently, the field in musculoskeletal (MSK) pain intervention has entered a renaissance. The MSK pain intervention is not restricted to joint injection any more but also includes fenestration of the tendons/ligaments, barbotage in calcific tendinitis, radiofrequency ablation of articular branches of joints, nonsurgical release of the nerve (e.g., carpal tunnel release), nerve release, and intraneural ablation.

As a result, there are a few books published in the arena of ultrasound-guided pain intervention. So, why did we decide to publish another one?

As our book title suggested, it is an illustrated procedural guide. We have 302 illustrations in 27 chapters. The generous numbers of illustration not just help the readers to grasp the concept of the anatomy and the procedure with ease; it also makes the learning enjoyable. We also make the layout easy and practical. A typical chapter started with an introduction of the procedure, the patient selection, and an overview of anatomy. Then, we presented the step-by-step ultrasound scanning procedure with illustrations. We also summarized all the clinical pearls from the expert. The chapter concluded with a brief review of the literature.

I am honored that three experience clinicians were willing to join me as the section editors: Dr. Anuj Bhatia for the peripheral structures, Dr. Rod Finlayson for the axial structures, and Dr. Sang-Hoon Lee for the musculoskeletal intervention. I sincerely thank them for the team effort. We are indebted to the expert contributors for the tireless effort to compose the chapters and invaluable input of their experience. Our hope is to provide clinicians interested in ultrasound-guided pain intervention an enjoyable learning experience and enrich them with the knowledge to benefit the patients suffering in pain.

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Montreal, QC, Canada — Roderick Finlayson
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About the Editor

Philip Peng is a full professor in the Department of Anesthesia and Pain Management of the University of Toronto and is currently the director of Anesthesia Pain Program in Toronto Western Hospital and interim director of Wasser Pain Management Center.

He has played an important role in the education of the pain medicine and established major teaching courses for pain in Canada such as the National Pain Refresher Course, Canadian Pain Interventional Course, and Ultrasound for Pain Medicine Course. The Royal College of Physicians and Surgeons of Canada (RCPSC) honored him with founder designation in pain medicine for his role in establishing pain medicine subspecialty in Canada. Besides, he currently serves as the chair of the Exam Committee in Pain Medicine in RCPSC and previously served as the chair of the Education Special Interest Group (SIG) of Canadian Pain Society and the founding executive of Pain Education SIG of International Association for the Study of Pain (IASP). He has been honored with numerous teaching awards at national and regional level.

Dr. Philip Peng is also a leader and pioneer in the application of ultrasound for pain medicine. Being one of the founding fathers for Ultrasound for Pain Medicine (USPM) SIG for ASRA (American Society of Regional Anesthesia), he was involved in the establishment of the guideline for Education and Training for USPM, which was adopted by five continents. He is the chair for the new Ultrasound for Pain Medicine Exam Certificate and chair for the Musculoskeletal Pain Ultrasound Cadaver workshop for ASRA and has been the chair or main organizer for various major teaching courses for USPM, including satellite meeting of the World Congress on Pain, International Pain Congress, combined Canadian and British Pain Society Conference, International Symposium of Ultrasound for Regional Anesthesia (ISURA), and Canadian Pain Interventional Course.

Furthermore, he has edited 7 books and published more than 150 peer-reviewed publications and book chapters.

Basic Principles and Physics of Ultrasound

1

Sherif Abbas and Philip Peng

Understanding the Physics of Ultrasound and Image Generation

Characteristic of Sound Wave

Audible sound wave lies within the range of 20–20,000 Hz. Ultrasound is a sound wave beyond audible range (>20,000 Hz). Ultrasound system commonly used in clinical settings incorporates transducers generating frequencies between 2 and 17 MHz. Some special ultrasound system even generates frequencies between 20 and 55 MHz. Sound waves do not exist in a vacuum, and propagation in gases is poor because the molecules are too widely spaced, which explains the use of gel couplant between the skin of the subject and the transducer interface to eliminate the air-filled gap.

Sound wave is a form of mechanical energy that travels through a conducting medium (e.g., body tissue) as a longitudinal wave producing alternating compression (high pressure) and rarefaction (low pressure) (Figs. 1.1 and 1.2). Sound propagation can be represented in a sinusoidal waveform with a characteristic pressure (P), wavelength (λ), frequency (f), period (T), and velocity. See Table 1.1 for details.

The speed of sound varies for different biological media, but the average value is assumed to be 1540 m/s for most human soft tissues. It can vary greatly, being as low as 330 m/s in air and as high as 4000 m/s through bone.

The wavelength (λ) is inversely related to the frequency (f). Thus, sound with a high frequency has a short wavelength and vice versa.

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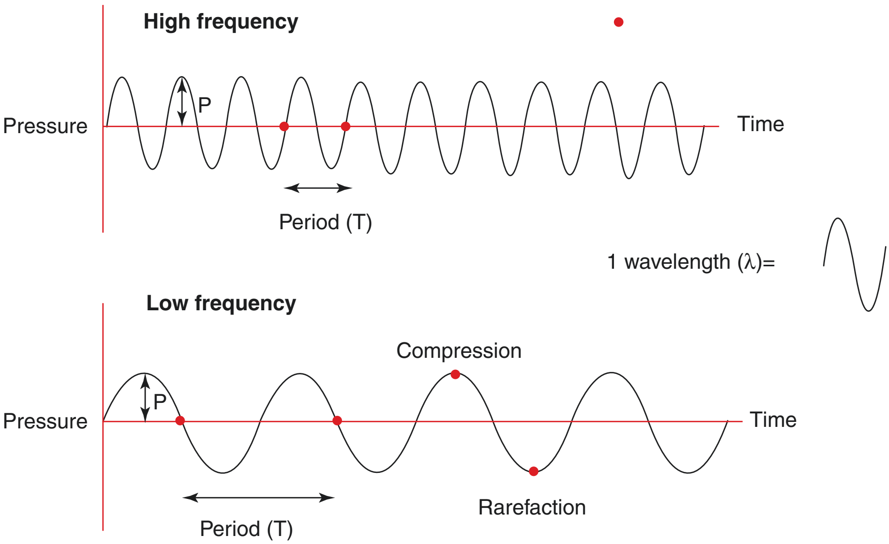

Fig. 1.1 Comparison of high-frequency and low-frequency waveform. (Reprinted with permission from Philip Peng Educational Series)

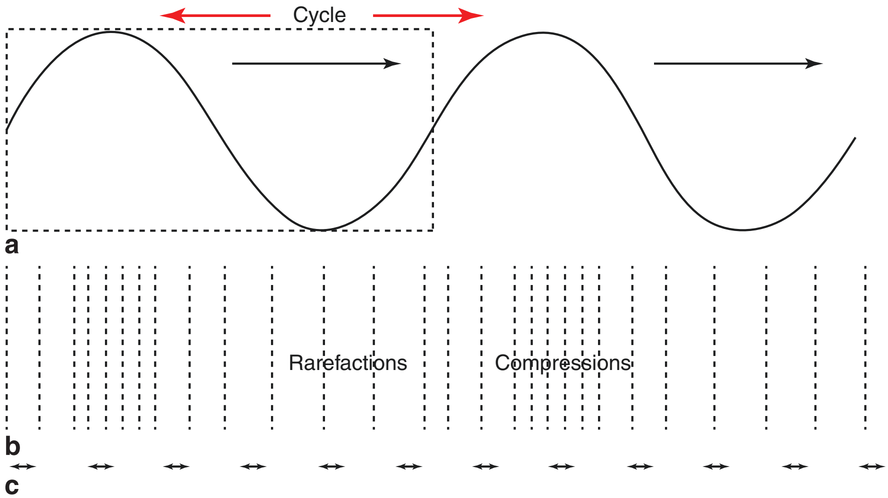

Fig. 1.2 A longitudinal wave showing alternating compression and rarefaction. (Reprinted with permission from Philip Peng Educational Series)

Generation of an Ultrasound Wave

An ultrasound wave is generated when an electric field is applied to an array of piezoelectric crystals located on the transducer surface. Electrical stimulation causes mechanical distortion of the crystals resulting in vibration and production of sound waves (i.e., mechanical energy). The conversion of electrical to mechanical

Table 1.1 Basic terminology

Terminology	Definition
Wavelength (λ)	The spatial period of the wave, and is determined by measuring the distance between two consecutive corresponding points of the same phase. It is expressed in meters (m)
Amplitude (A)	A measure of the height of the wave, i.e., maximal particle displacement. It is expressed in meters (m)
Period (T)	The time taken for one complete wave cycle to occur. The unit of period is seconds (s)
Frequency (f)	The number of completed cycles per second. Thus, it is the inverse of the period (T) of a wave. The unit of frequency is hertz (Hz). Medical imaging uses high-frequency waves (1–20 MHz)
Velocity (c)	The speed of propagation of a sound wave through a medium (m/s). It is the product of its frequency (f) and wavelength (λ)
Energy (E)	The energy of a sound wave is proportional to the square of its amplitude (A). This means that as the amplitude of a wave decreases (such as with deeper penetration), the energy carried by the wave reduces drastically
Power (P)	Defines as the energy (E) delivered per unit time (t)

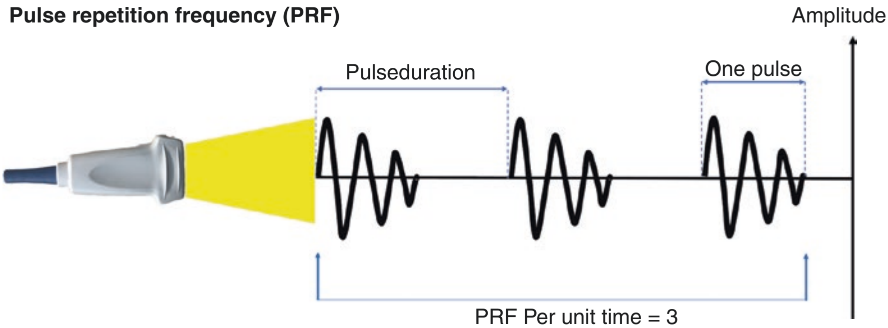

Fig. 1.3 Pulse repetition frequency. (Reprinted with permission from Philip Peng Educational Series)

(sound) energy is called the converse piezoelectric effect. Each piezoelectric crystal produces an ultrasound wave. The summation of all waves generated by the piezoelectric crystals forms the ultrasound beam. Ultrasound waves are generated in pulses (intermittent trains of pressure waves), and each pulse commonly consists of two or three sound cycles of the same frequency.

The pulse length (PL) is the distance traveled per pulse. Waves of short pulse lengths improve axial resolution for ultrasound imaging. The PL cannot be reduced to less than 2 or 3 sound cycles by the damping materials within the transducer.

Pulse repetition frequency (PRF) is the rate of pulses emitted by the transducer (number of pulses per unit time) (Fig. 1.3). Ultrasound pulses must be spaced with enough time between pulses to permit the sound to reach the target of interest and return to the transducer before the next pulse is generated. The PRF for medical imaging ranges from 1 to 10 kHz. For example, if the PRF = 5 kHz and the time between pulses is 0.2 ms, it will take 0.1 ms to reach the target and 0.1 ms to return to the transducer. This means the pulse will travel 15.4 cm before the next pulse is emitted (1540 m/s × 0.1 ms = 0.154 m in 0.1 ms = 15.4 cm).

Generation of an Ultrasound Image

An ultrasound image is generated when the pulse wave emitted from the transducer is transmitted into the body, reflected off the tissue interface, and returned to the transducer. The schematic diagram above showed the transducer waits to receive the returning wave (i.e., echo) after each pulsed wave (Fig. 1.4). The transducer transforms the echo (mechanical energy) into an electrical signal which is processed and displayed as an image on the screen. The conversion of sound to electrical energy is called the piezoelectric effect.

The image can be displayed in a number of modes (Fig. 1.5):

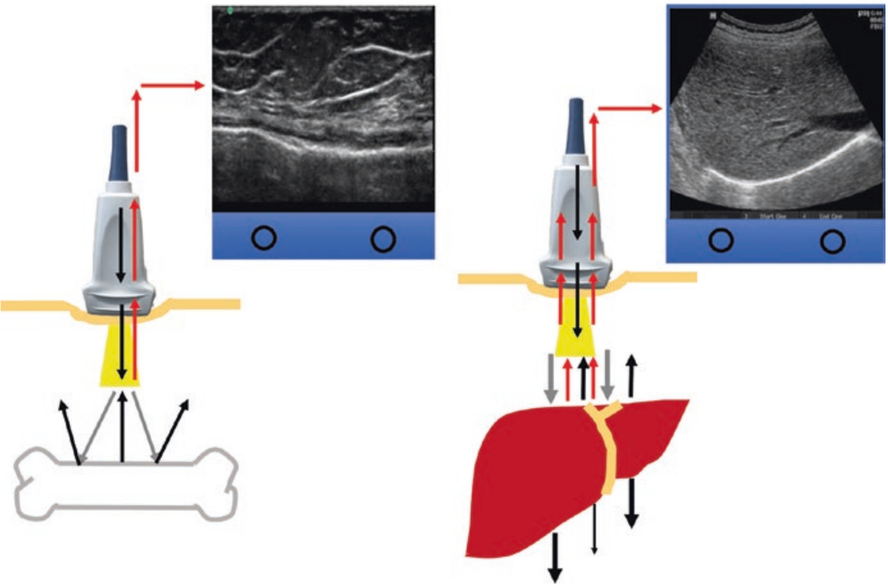

Fig. 1.4 Ultrasound wave interaction with body tissue. (Reprinted with permission from Philip Peng Educational Series)

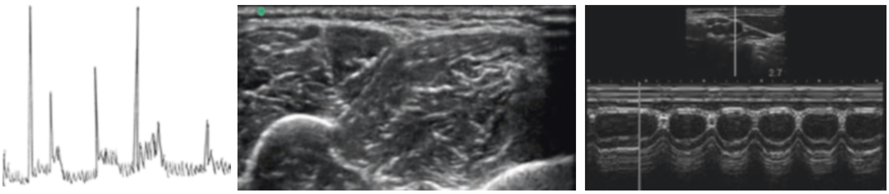

Fig. 1.5 Three different modes of ultrasound. (Reprinted with permission from Philip Peng Educational Series)

- *Amplitude (A)* mode is the display of amplitude spikes in the vertical axis and the time required for the return of the echo in the horizontal axis.
- *Brightness (B)* displays a two-dimensional map of the data acquired and is most commonly used for ultrasound guided intervention.
- *Motion (M)* mode, also called time motion or TM mode, displays a one-dimensional image usually used for analyzing moving body parts. This mode records the amplitude and rate of motion in real time and is commonly used in cardiovascular imaging.

Ultrasound Tissue Interaction

As the ultrasound beam travels through tissue layers, the amplitude of the original signal becomes attenuated as the depth of penetration increases (Fig. 1.6).

Attenuation (energy loss) is due to:

1. Absorption (conversion of acoustic energy to heat)
2. Reflection
3. Scattering at interfaces

In soft tissue, 80% of the attenuation of the sound wave is caused by absorption resulting in heat production. Attenuation is measured in decibels per centimeter of

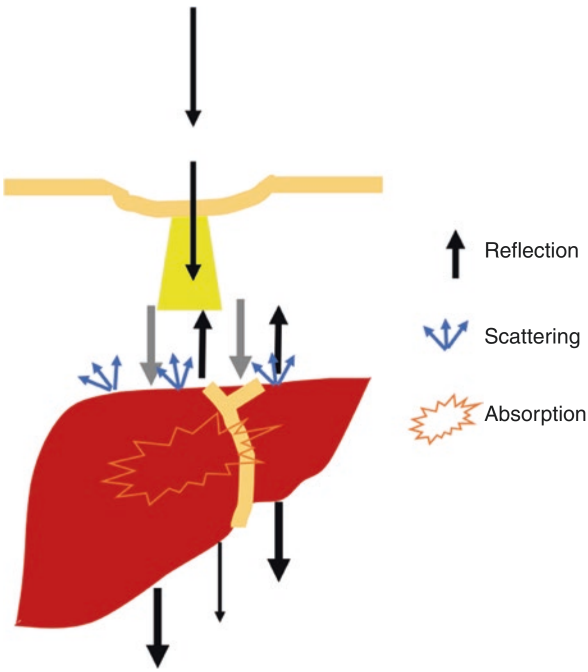

Fig. 1.6 Different types of attenuation. (Reprinted with permission from Philip Peng Educational Series)

tissue and is represented by the attenuation coefficient of the specific tissue type. The higher the attenuation coefficient, the more attenuated the ultrasound wave is by the specified tissue.

Absorption

Absorption is the process of transfer of the ultrasound beam's energy to the medium through which it travels through heat generation and it accounts for most of the wave attenuation. The quality of the returning sound waves depends on the attenuation coefficient of different tissue.

The degree of attenuation also varies directly with the frequency of the ultrasound wave and the distance traveled (Fig. 1.7 and Table 1.2). Generally speaking,

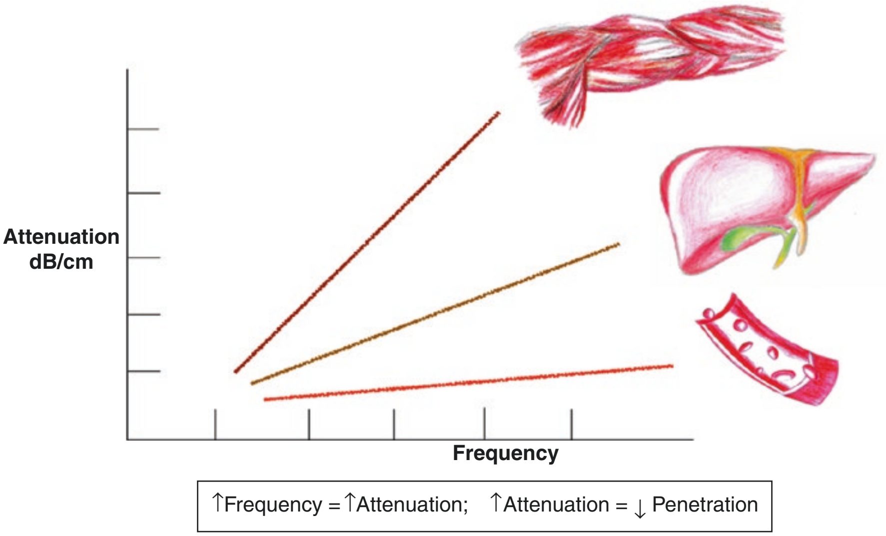

Fig. 1.7 Variation of attenuation with frequency in different organs. (Reprinted with permission from Philip Peng Educational Series)

Table 1.2 Attenuation coefficient of various tissues

Material	α (dB/cm)
Blood	0.18
Fat	0.6
Muscle (across fibers)	3.3
Muscle (along fibers)	1.2
Aqueous and vitreous humor of the eye	0.1
Lens of the eye	2.0
Skull bone	20
Lungs	40
Liver	0.9
Brain	0.85
Kidney	1.0
Spinal cord	1.0
Water	0.0022
Castor oil	0.95
Lucite	2.0

a high-frequency wave is associated with high attenuation, thus limiting tissue penetration, whereas a low-frequency wave is associated with low tissue attenuation and deep tissue penetration.

To compensate for attenuation, it is possible to amplify the signal intensity of the returning echo. The degree of receiver amplification is called the gain. Increasing the gain will amplify only the returning signal and not the transmit signal. An increase in the overall gain will increase brightness of the entire image, including the background noise. Preferably, the time gain compensation (TGC) is adjusted to selectively amplify the weaker signals returning from deeper structures.

Reflection

Attenuation also results from reflection and scattering of the ultrasound wave. The extent of reflection is determined by the difference in acoustic impedances of the two tissues at the interface, i.e., the degree of impedance mismatch.

Acoustic impedance is the resistance of a tissue to the passage of ultrasound. The higher the degree of impedance mismatch, the greater the amount of reflection (Table 1.3). The degree of reflection is high for air because air has an extremely low acoustic impedance (0.0004) relative to other body tissues. The bone also produces a strong reflection because its acoustic impedance is extremely high (7.8) relative to other body tissues. For this reason, it is clinically important to apply sufficient conducting gel (an acoustic coupling medium) on the transducer surface to eliminate any air pockets between the transducer and skin surface. Otherwise, much of the ultrasound waves will be reflected limiting tissue penetration.

The angle of the incidence is also a major determinant of reflection. An ultrasound wave hitting a smooth mirror-like interface at a 90° angle will result in a perpendicular reflection. An incident wave hitting the interface at an angle <90° will result in the wave being deflected away from the transducer at an angle equal to the angle of incidence but in the opposite direction (angle of reflection). When this happens, the signal of the returning echo is weakened, and a darker image is displayed (Fig. 1.8). This explains why it is difficult to visualize a needle inserted at a steep angle (>45° to the skin surface).

Specular reflection occurs at flat, smooth interfaces where the transmitted wave is reflected in a single direction depending on the angle of incidence. Examples of specular reflectors are fascial sheaths, the diaphragm, and walls of major vessels

Table 1.3 Acoustic impendence of various tissues

Body tissue	Acoustic impedance (10^6 RayIs)
Air	0.0004
Lung	0.18
Fat	1.34
Liver	1.65
Blood	1.65
Kidney	1.63
Muscle	1.71
Bone	7.8

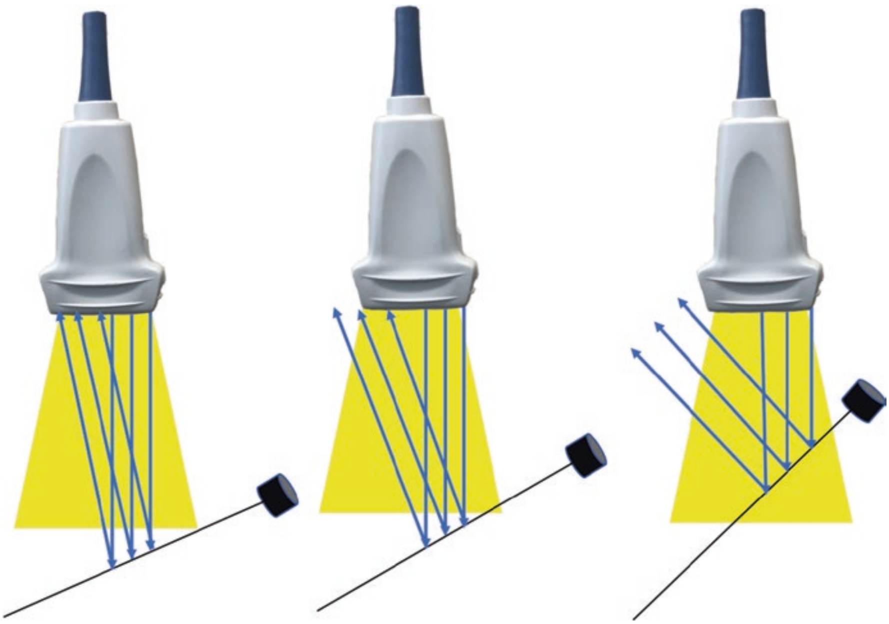

Fig. 1.8 Angle of incidence. (Reprinted with permission from Philip Peng Educational Series)

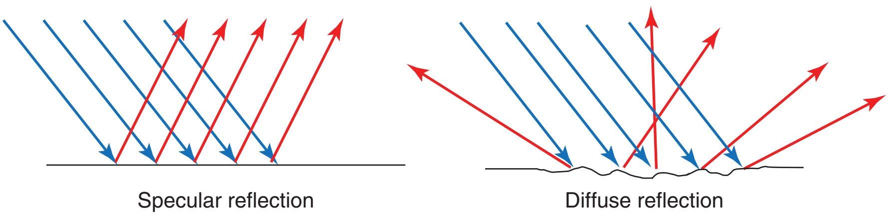

Fig. 1.9 Different types of reflection. (Reprinted with permission from Philip Peng Educational Series)

(Fig. 1.9). Block needles are also strong specular reflectors. For specular reflection to occur, the wavelength of the ultrasound wave must be smaller than the reflective structure. Otherwise, scattering will occur.

Scattering

Reflection in biological tissues is not always specular. Scattering (diffuse reflection) occurs when the incident wave encounters an interface that is not perfectly smooth (e.g., surface of visceral organs). Echoes from diffuse reflectors are generally weaker than those returning from specular reflectors. Scattering also occurs when the wavelength of the ultrasound wave is larger than the dimensions of the reflective structure (e.g., red blood cells). The reflected echo scatters in many different

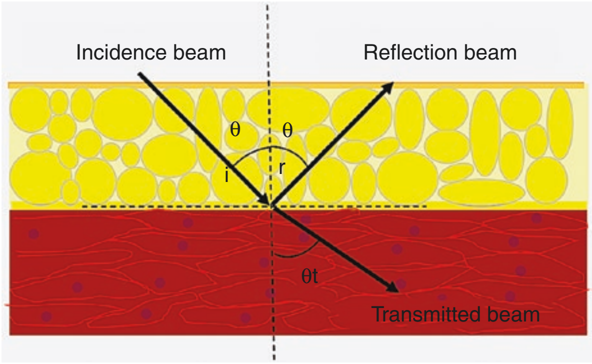

Fig. 1.10 Transmission of beam. (Reprinted with permission from Philip Peng Educational Series)

directions resulting in echoes of similar weak amplitudes. Ultrasonic scattering gives rise to much of the diagnostic information we observe in medical ultrasound imaging.

Transmission

After reflection and scattering, the remainder of the incident beam is refracted with a change in the direction of the transmitted beam (Fig. 1.10). Refraction occurs only when the speeds of sound are different on each side of the tissue interface. The degree of beam change (bending) is dependent on the change in the speed of sound traveling from one medium on the incident side to another medium on the transmitted side (Snell's Law). With medical imaging, fat causes considerable refraction and image distortion, which contributes to some of the difficulties encountered in obese patients. Refraction encountered with bone imaging is even more significant leading to a major change in the direction of the incident beam and image distortion.

The final image on the screen of an ultrasound machine is the result of the interaction of ultrasound waves with the tissues being examined. As the ultrasound wave travels through the tissues, it loses amplitude, and hence energy (attenuation), which is the summative effect of absorption, reflection, and refraction of ultrasound waves.

Image Acquisition and Processing

Transducer Basic

An ultrasound transducer has a dual functionality. It is responsible for both the production of ultrasound waves and, after a set period of time, the reception of waves reflected from the tissues. This is called pulsed ultrasound. The pulse repetition frequency (PRF) is the number of pulses emitted by the transducer per unit of time. The PRF for medical imaging devices ranges from 1 to 10 kHz

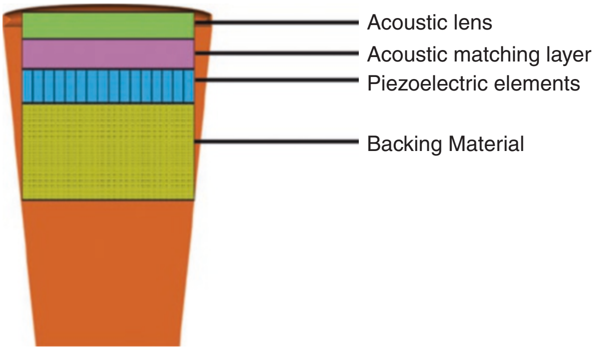

Fig. 1.11 Anatomy of a transducer. (Reprinted with permission from Philip Peng Educational Series)

The ultrasound transducer has the following layers (Fig. 1.11):

(a) Backing material: located behind the piezoelectric element, it serves to prevent excessive vibration. This causes the element to generate ultrasonic waves with a shorter pulse length, improving axial resolution in images.
(b) Piezoelectric elements: they generate ultrasonic waves and also generate images. Piezoelectric ceramic (PZT: lead zirconate titanate) is most commonly used because of their high conversion efficiency.
(c) Acoustic matching layer: this reduces the acoustic impedance mismatch between the transducer and the object and thus minimizes reflection off the interface. This is usually made up of a resin.
(d) Acoustic lens: the acoustic lens prevents the ultrasonic waves from spreading and focuses them in the slice direction to improve the resolution.

Transducer Selection

The choice of which transducer should be used depends on the depth of the structure being imaged. The higher the frequency of the transducer crystal, the less penetration it has but the better the resolution. Therefore, if more penetration is required, you need to use a lower-frequency transducer with the sacrifice of some resolution.

Transducer characteristics, such as frequency and shape, determine ultrasound image quality. For simplicity, there are three types of transducers.

The linear transducer frequencies used for superficial structures and peripheral nerve blocks range from 6 to 15 MHz (Fig. 1.12). Curvilinear (or curved) transducers are most useful for deeper structures or imaging requiring a wide field of view (e.g., spine). Microconvex is commonly used for small acoustic window such as cardiac scanning.

For superficial structures (e.g., nerves in the interscalene region), it is ideal to use high-frequency transducers in the range of 10–15 MHz, but depth of penetration is often limited to 2–3 cm below the skin surface (Fig. 1.13). For visualization of

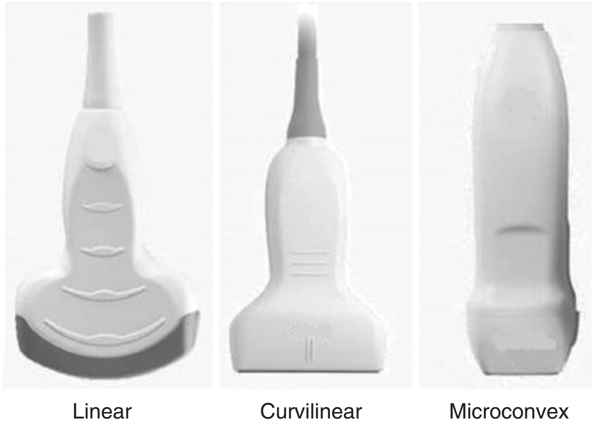

Fig. 1.12 Different types of transducer. (Reprinted with permission from Philip Peng Educational Series)

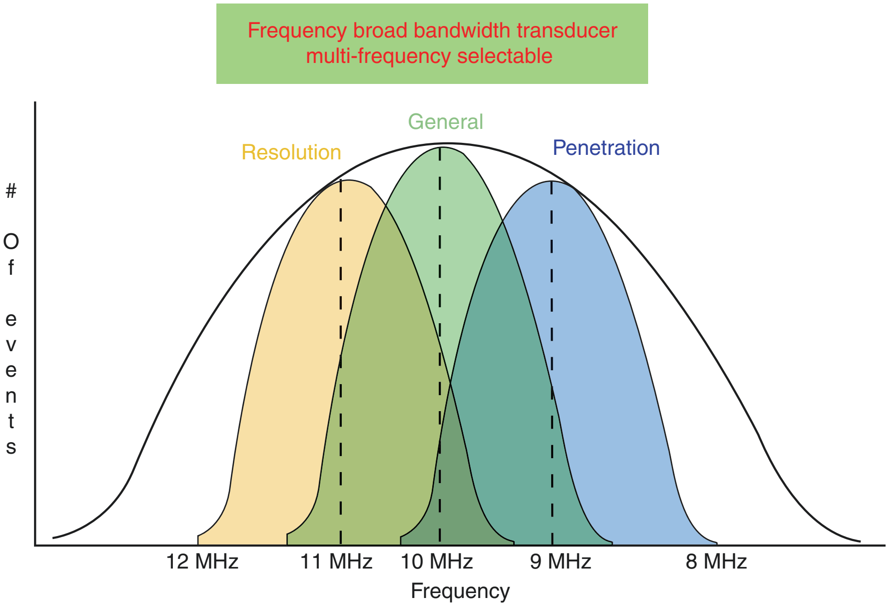

Fig. 1.13 Frequency, resolution, and penetration. (Reprinted with permission from Philip Peng Educational Series)

deeper structures (e.g., in the gluteal region) or when a wider field of view is required, it may be necessary to use a lower-frequency transducer (2–5 MHz) because it offers ultrasound penetration of 4–5 cm or more below the skin surface. However, the image resolution is often inferior to that obtained with a higher-frequency transducer.

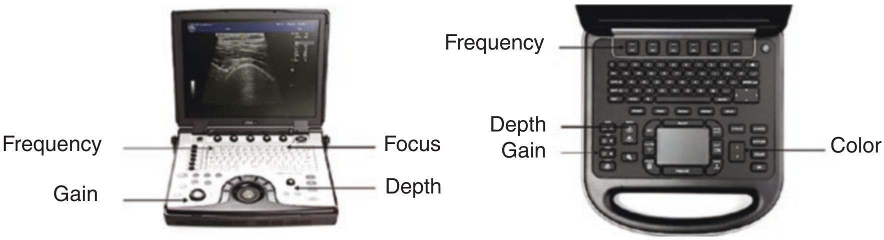

Fig. 1.14 Basic button of ultrasound machine. (Reprinted with permission from Philip Peng Educational Series)

Basic Operation of Ultrasound Machine

To obtain a good ultrasound picture, clinician should be familiar with the basic function and buttons in the ultrasound machine: gain, depth, color (Doppler), and focus (Fig. 1.14).

Gain

The echo signals returning from the tissues reach the crystals and produce an electric current (piezoelectric effect). This is then converted to a pixel on the image, with the energy of the returning wave being proportional to the brightness of the pixel dot on the image. The amplitude of the returning echo signals is very small to be properly displayed on a screen. Hence, it needs amplification. Amplification of signal can be adjusted using the GAIN button, but the use of it adds to "background noise."

Time gain compensation It is the preferential enhancement of signals at different depths returning from deeper tissues (Fig. 1.15). Thus, the echoes returning from similar reflectors can be represented by the same shade of gray regardless of their depth.

Depth

The depth of the field should be adjusted to the area of interest (Fig. 1.16). Too shallow may miss the important information from the background in the deep field, and too deep will diminish the quality of the image in the superficial field.

Focus

The shape of the beam varies and is different for each transducer frequency. There is a fixed focused region of the ultrasound beam which is indicated on the system with a small triangle to the right of the image. This indicates the focal zone of that

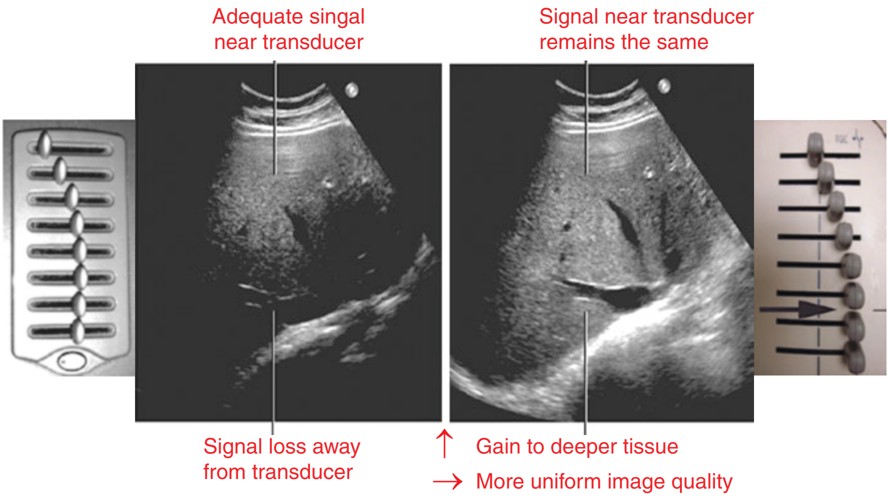

Fig. 1.15 Time gain compensation. (Reprinted with permission from Philip Peng Educational Series)

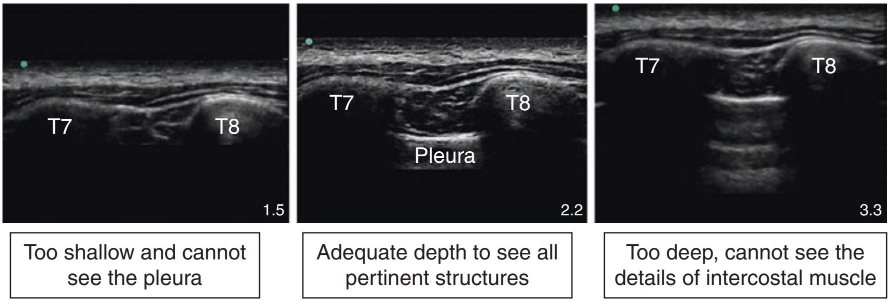

Fig. 1.16 Depth and image acquired. (Reprinted with permission from Philip Peng Educational Series)

transducer and is where the best resolution can be achieved with that particular transducer (Fig. 1.17). Effort should be taken to position the object of interest in the subject to within that focused area to obtain the best detail by adjusting the FOCUS button (Fig. 1.18).

Doppler

The Doppler information is displayed graphically using spectral Doppler, or as an image using color Doppler (directional Doppler) or power Doppler (non-directional Doppler).